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Electro-Optical Monitoring and Analysis of Human Cognitive Processes

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1. Introduction

Miniaturization of biomedical sensors has increased the importance of microsystem technology in medical applications, particularly microelectronics and micromachining. Microelectronics enables a mass-production of planar microsensors with high reproducibility, low cost, advanced performance, and built-in smartness. Thin-film interdigital arrays (IDA) of microelectrodes are among the most commonly used periodic structures in a wide variety of sensors and transducers (Mamishev et al., 2004).
This chapter presents a new approach to biomedical monitoring and analysis of selected human cognitive processes. The proposed method measures human skin conductivity and heart rate using a dedicated monitoring system with the developed IDA microelectrodes (Sheppard, 1993) as well as skin temperature and reflectance using an optical spectrometer, which offers continuous monitoring and analysis of different electrophysiological aspects of human physiology in a completely safe and non-invasive manner. This technique also has no undesired influence on natural physiological processes. The main goal is monitoring of the psycho-galvanic reflex (PGR) of the human skin that might be very useful for the identification of psychical stress effect that is performed in different medical as well as psychological experiments.
A portable monitoring system, applicable also in wireless measurement environment, was designed and developed. Comparison to a standard laboratory-like bridge-based measurement system was done in terms of accuracy, sensitivity and other main features. The proposed mobile monitoring equipment utilizes microprocessors with an RF wireless communication modules used for data transfer between the measurement modules and a personal computer. A graphical user interface (GUI), developed in C++ under Windows XP platform, has been developed too, and is used to provide necessary calibration of the measurement as well as storage, displaying and postprocessing of the measured data (real and imaginary components of the skin impedance as well as phase of the measured impedance). The measurement method itself and the achieved results are discussed in the bellow sections.

The explosive growth in recent wireless communications has caused increased demands for wireless products that would be low-cost, low-power, and compact in size. Moreover, extensive research and development in semiconductor industry in latest years has led towards the novel technologies, and consequently, new products such as smart systems, system-on-chip (SoC), etc. have been widely developed and used. There are modern high-performance microcontrollers with many useful features (A/D converters, several types of interface, RF wireless transceivers, etc.) available and enabling the realization of modern compact and low-power portable measurement systems. Similarly, massive expansion in electronics trade has enabled to affiliate electronics to diverse domains like the health care that affords new opportunities. Different electronic-medical measurement systems and equipments are indispensable parts of various surgery procedures and help diagnostic a lot of affections. Therefore, biomedical monitoring can be very useful in the health care and psychology, and moreover, it can enhance the quality of life in areas of relaxing physiotherapy or professional and leisure sport activities. Even though measurements of the selected physiological parameters e.g. electrical conductivity (impedance) of the human skin surface and skin temperature have a long history, the way how physiological changes in the human tissue are reflected in electrical impedance has not been very cleared up yet (Olmar, 1998), (Wakelam, 2000).

The proposed biomedical monitoring approach employs advanced wireless technologies in controlling the measurement setup and transferring the measured data to a personal computer. There are many benefits of the proposed portable solution, such as wireless monitoring of the tested person, accurate and sensitive measurement, free movement of the person being tested, real life monitoring, no side effects (e.g. additional stress due to the fact that the person is in the laboratory conditions and become aware of being tested), etc. Furthermore, it can help to identify abnormal changes in human physiological and psycho-physiological reactions under certain psychical stress stimuli. This allows reliable diagnostics of the undesired stress that might be a very negative factor, influencing not only human performance but causing also serious health problems.

2. Stress and Physiological Processes

2.1 Stress Phenomena

What the psychical stress really is and may cause? Stress is a very troublesome and undesired factor dramatically affecting the central neural system and it might invoke significant psychical as well as health problems and inconveniences (Gerasimov, 2003). According to Canadian psychologist Hans Selye, stress is a physiological response of the body to certain physical demands. Such demands are known as stressors. There are three categories of psychological stressors:

- First category comprises stressors that cause frustration. We experience frustration when being blocked from reaching a goal. Our degree of psycho-physiological reactivity to frustrating situations may be affected by heredity.
- Second category of stressors comprises those that cause tension. We experience tension when we must fulfill responsibilities that tax our abilities.
- Third category of stressors comprises those that cause conflicts when we are torn by two or more potentional courses of actions.

Though major life changes are important stress loads, including events, they are not sole once. We can mention other social stress rating scales: from death of spouse (100), followed by marriage (50), change in family members (44), change of responsibility in work (29), change sleeping habits (19), vacations (15), Christmas (13), etc.
Generally, stress is defined by so called General Adaptation Syndrome. Author (Selye, 1976) believes that the syndrome represents body´s defence against the stress. Selye argues that the initial symptoms of almost any disease or trauma are virtually identical, that is, the body responds in the same way to any source of stress, whether it is external and environmental or whether it arises from within the body itself. He has defined stress as „the individual´s psycho-physiological response mediated largely by the autonomic nervous system and the endocrine system, to any demands made on the individual".
Stress has been related to a number of deleterious and costly individual problems (e.g., headaches, gastrointestinal disorders, anxiety, hypertension, coronary heart disease, depression) and organizational outcomes (e.g. job dissatisfaction, burnout, accidents, loss of productivity, absenteeism, turnover) (Ganster & Schaubroeck, 1991), (Gupta & Beehr, 1979), (Ironson, 1992), (Murphy et. al, 1995), (Quick et. al, 1987), (Sauter&Murphy, 1995). These consequences have sustained a continuing interest in assessing stress at work in order to understand the etiology of workplace stress, pinpoint sources of stress, and guide the use of stress reduction interventions. Current techniques for measuring stress fall into one of three categories: self - reports, behavioural and cognitive functions measures, and medical/biological measures (where our study is oriented). To facilitate non-invasive field and laboratory research, the present study has focused on the development of a self-reporting measure of work stress that can be administered rapidly and uniformly, using a structured, closed-ended response format. In developing this Stress in General (SIG) measure, we took a broad approach that avoided links to specific stressors or strains, in contrast to the many previous efforts to measure the work stress that counted on cataloguing the presence of various classes of stressors or short-term strains. The need for the present validation study is underscored by the existence of published studies that have used early prepublication versions of the scale. The instrument is clearly popular and practical even lacking published validity data.
Thus, monitoring of some high-risk groups of patients enables the effective prevention and remarkable reduction of possible health risks associated with the chronic consequences of the stress factors' influence. Stress monitoring might be very helpful also in a wide range of psychological applications, such as clinic psychology, treatment of drug-dependent people, monitoring of important and high-reliability jobs (dispatchers, pilots, drivers, etc.) and others. Thus, measurement of selected physiological variables offers experts a possibility to observe and analyze complex psycho-physiological processes that might considerably contribute to the optimalization of diagnostic and therapeutic procedures.

2.2 Psychogalvanic Reflex

Recently, an increased accuracy of biomedical experimental techniques brings great interest in the field of psycho-physiological correlatives. In general, it has been observed that psycho-galvanic reflex represents psycho-physiological activation, starting from the lowest amplitude in sleep up to the top amplitude under a strong activation. The amplitude fluctuation depends on the level of psychological activation, where the skin conductivity represents volume of sympathetic activity. It has been proven that so called psycho-galvanic reflex (a change in the human skin conductivity under stress influence), sensed continuously within a given time period, offers satisfactory information for stress, overwhelmed excitement, or a shock identification. From the accuracy and sensitivity points of view, the skin conductivity parameters are best sensed using microelectrodes (as explained bellow). The thermal sense represents one of the best characteristics of physical relaxation, where temperature changes are inflicted due to increased amount of blood flowing in the bloodstream and due to vasodilatation (Brezina, 2007).

Although, technical realization of these measurements might be very simple, in practice, there is a problem with measurement reproducibility and comparison. First, it was assumed that increase in the skin conductivity during a stress stimulus is only caused by the skin perspiration. Later, a very important factor of the *potential barrier* near the *stratum lucidum* layer, which thickness changes due to the nervous system, was discovered and proven (Olmar, 1998), (Weis et. al, 1995), (Shepherd, 2007), (Qubit systems, 2004).

If different electrodes are applied on human skin, various space distribution of electrical field into the skin can occurs.

In case of using macroelectrodes, when the distance between the coupled electrodes is greater than the thickness of electric active layers of skin h (*stratum corneum* (the outermost layer of the skin) with potential barrier) d >> h, the vector intensity lines of the electric field are enclosed perpendicular to the skin surface across the planar skin structures through *dermis* with high conductance (Fig. 1 a, b).

If microelectrode pairs are utilized, when the distance between the electrodes is less than thickness of electric active layers of skin: d < h > s (s – thickness of *stratum corneum*), then the lines of electric field are enclosed in parallel direction relative to laminar skin structures of epidermis (of lower conductance) (in *stratum corneum*). From inner layers of skin, the electric field intensity lines are embossed to the surface (to the area with a lower conductivity) by the influence of the potential barrier which is generated by electrical double-layer around *stratum lucidum* (Fig.1c). Dynamic electrical properties of potential barrier reflect fact that it is responsible for trans-epidermal transports – substance exchange, water transport and thermo regulation (Fig. 1c). Under a stress stimulus the potential barrier narrows down and the electric field can reach inner layers of human skin with higher conductivity, and therefore, the total conductivity is increases (Fig. 1d). Such configuration is therefore ideal for the analysis of electrophysiological processes in human skin under stress (Vavrinsky et. al, 2008).

In case "too small" microelectrodes the vector intensity lines of the electric field are enclosed in top layers of *stratum corneum* and the flow of electric lines is independent on thickness of potential barrier (Fig. 1e, f). Such electrodes are more ideal for surface analysis in cosmetic.

The results of analytical analysis also showed that in case of non-symmetric coplanar electrodes (Fig. 2), the electric field is more enclosed in the outer layers of the skin laminar structures (stratus corneum). This system consists of the periodical electrode structure with different sizes. In a non-symmetric structure, the density of the electric field intensity lines along the planar structures of the skin is 30 % higher in outer layers (Ivanic et. al, 2003).

The greatest degree of conductivity change occurs in the skin of palms and bottom parts of fingers, but because of practical reasons, we put our microelectrodes on the second best sensitive part of human body – wrist bottom of non-dominant hand (left for right-handers) (Weis et. al, 1995), (Wakelam, 2000).

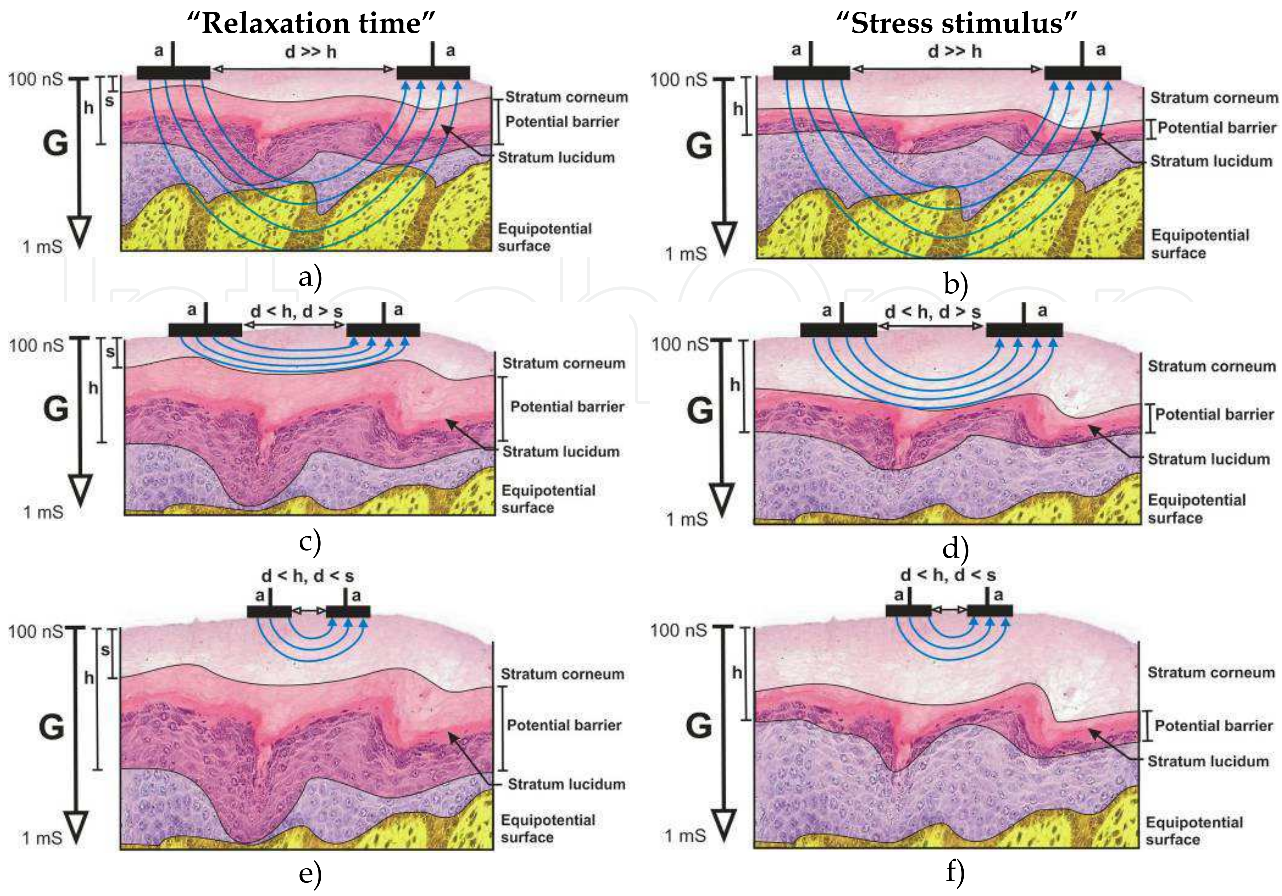


Fig. 1. The dominant vector intensity lines of the electric field in human skin:
- macroelectrodes: a) in relaxation time, b) under stress stimulus
- microelectrodes: c) in relaxation time, d) under stress stimulus
- "too small" microelectrodes: e) in relaxation time, f) under stress stimulus

2.3 Developed Microelectrodes

For non-invasive biomedical monitoring of psycho-physiological processes based on skin conductivity measurements, four types of IDA microelectrodes with the following configuration and sizes have been developed, produced and used:

- Non-symmetric configuration
  - 15 μm/25 μm /50 μm (finger/gap/finger) (Fig. 2)
- Symmetric configuration
  - 100 μm/100 μm (finger/gap dimensions)
  - 200 μm/200 μm
  - 400 μm/400 μm

The total size of the microelectrode chips is 10 mm x 15 mm. The microelectrodes were made from Pt or Au thin film to minimize the polarization effect (Fig. 2). The microelectrodes were fabricated by a standard thin film technology: Pt (Au) films (150 nm in thickness) underlaid by Ti film (50 nm) were deposited by rf sputtering on Al_2O_3 substrates and microelectrodes were lithographically patterned by lift-off technique.

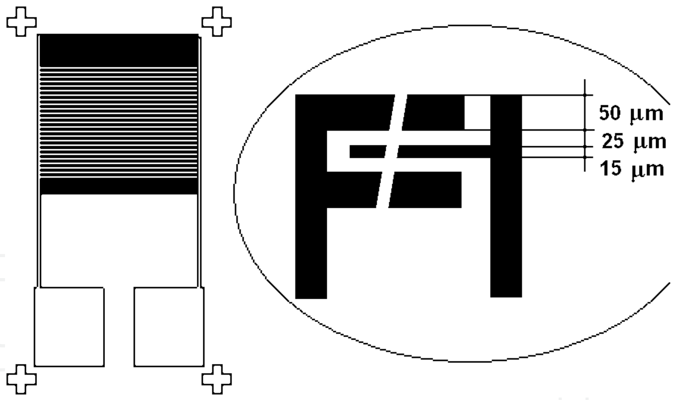


Fig. 2 Layout of the non-symmetric electrode chip

2.4 Experimental Skin Conductivity Measurements

The electrodermal response (EDR) to stress stimulus was in first experiments detected by variations in the skin (ΔG) using classical laboratory measurement equipment (Fig. 3).

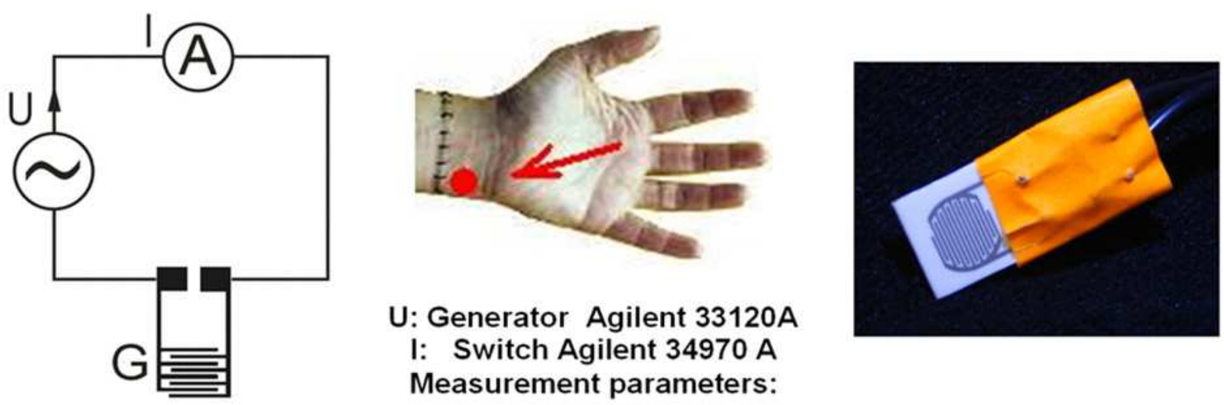


Fig. 3. Electrical laboratory equipment

During conductivity measurements, a drift of the output signals occurs due to polarization effects in the human skin – electrodermal phenomenon (EDF). We investigated the drift of output signals due to EDF and sweat hydration of the skin outer layers (stratus corneum and lucidum) in the time domain. These experiments were done using 200 µm/200 µm microelectrodes for dry skin (Fig. 4a) and for wet (sweaty) skin (Fig. 4b). Measured curves were interpolated by the exponential function $G(t) = A + B(1-e^{-t/C})$, where G is the skin conductivity, t is time and A, B, C are constants. This function is considered to be very suitable for description of polarization and hydration effects in the human skin. The performed experiments also showed that the signal stabilization (steady state) occurs sooner for hydrated skin (10 - 40 minutes) than for dry skin (40 minutes and more). For typical macroelectrodes, the increase in the conductivity due to the EDF effect is ended in 30 - 40 minutes (Weis et. al, 1995). During the several initial seconds, minutes of measurements it is possible to minimize and compensate the undesired output signal drift (caused by EDF, Fig. 5a)) by means of a proper software (Fig. 5b)). The software analysis program for the skin conductance activity was developed in the development environment Agilent VEE.

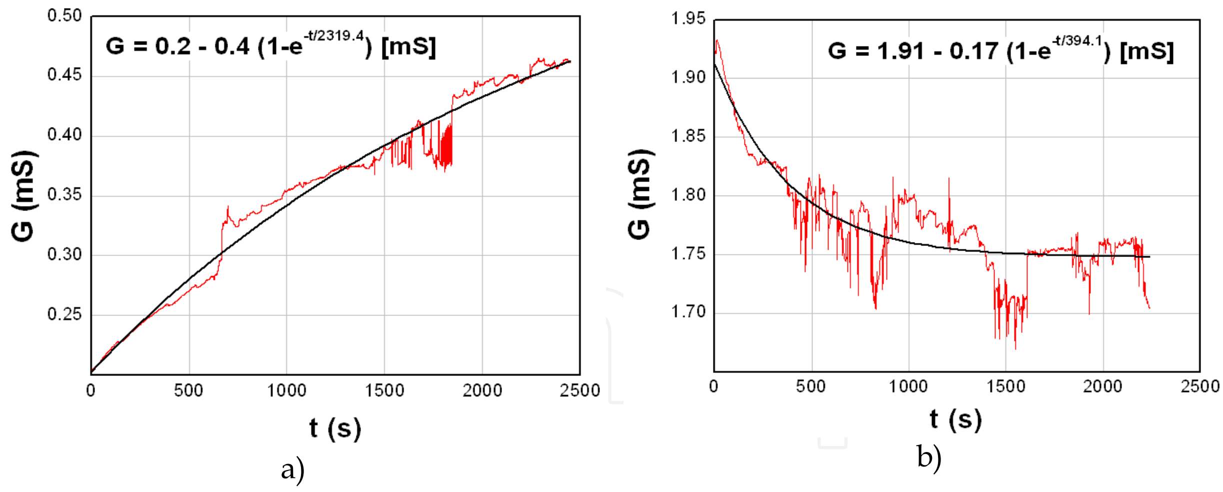
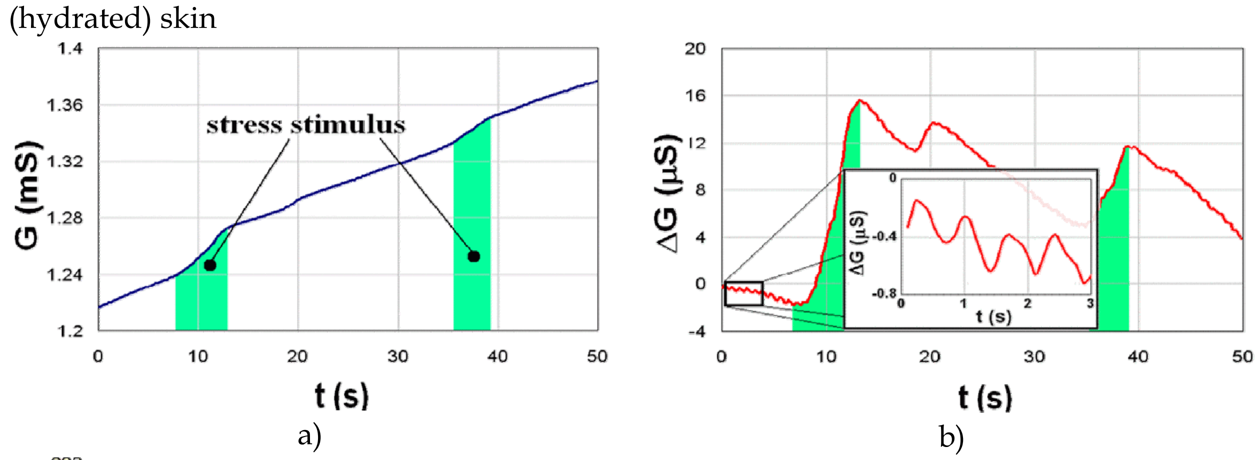
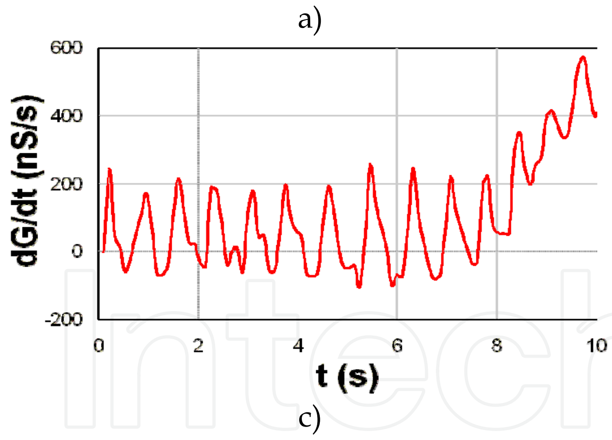


Fig. 4. Electrodermal phenomenon and effect of sweat hydration: a) dry skin, b) sweat (hydrated) skin





$$y = \sum_{t=0}^{N} \left\{ \left(BPF[\![FFT[x(t)]]\!]_{f_{Low_1}}^{f_{High_1}} \right) * \right.$$

$$\left. * \left(BPF \left[\!\left[FFT\left[\frac{dx(t)}{dt}\right]\right]\!\right]_{f_{Low_2}}^{f_{High_2}} - A \right) \right\}$$

d)

Fig. 5. Typical time dependences of EDR a) uncorrected, b) corrected EDR and its zoom comprising the heart pulse, c) derived uncorrected signal comprising the heart pulse, d) heart pulse analysing formula

These experiments led to a very important result: the developed microelectrode probes are able to monitor electrodermal response as well as heart pulse simultaneously (Fig. 5b). The heart pulse was observable by derivation of the measured signal (Fig. 5c). For this purpose, dedicated software, based on the designed formula (Fig. 5d), has been used. In this formula, *BPF* stands for a band-pass filter in the frequency range from f_{LOW} to f_{HIGH}, *FFT* represents Fast Fourier Transform, and *A* is a constant. For our experiments, the following ideal set of parameters has been found: $f_{Low1} = f_{Low2}$ = 0,5 Hz, $f_{High1} = f_{High2}$ = 3 Hz (heart-beat of 30-180 min^{-1}), N = 5, and A = 0,25. This software is under further development.

In the next experiments, the influence of microelectrodes size (symmetric: 200 μm/200 μm, 100 μm/ 100 μm, non-symmetric 15 μm/25 μm/50 μm) on the output signals were investigated. As mentioned above, the different types of microelectrodes generate the electrical field enclosed in various layers of the skin. Results were obtained using input signal amplitude and frequency of 3 V and 1 kHz, respectively, while employing the EDF correction. It is shown (Fig. 6), that the output signal measured by the non-symmetric IDA electrodes is very low in comparison to the results obtained by the symmetric IDA 100 μm/ 100 μm and 200 μm/200 μm microelectrode arrangement. It is inflicted by the fact that the penetration depth of the electric field generated by the non-symmetric microelectrodes is insufficient to reach the potential barrier of the skin, which is very sensitive to the detection of psycho-physiological processes like the psychical stress (Weis et. al, 1995), (Vavrinsky et. al, 2008). In case of the non-symmetric microelectrodes, most of the electrical field intensity lines (about 80 %) are enclosed in the depth of 0 - 25 μm , while the most important and sensitive layer of the potential barrier is placed more than 30 μm beneath the surface. The low response for the sensor is due to sweat in upper layer - stratum corneum.

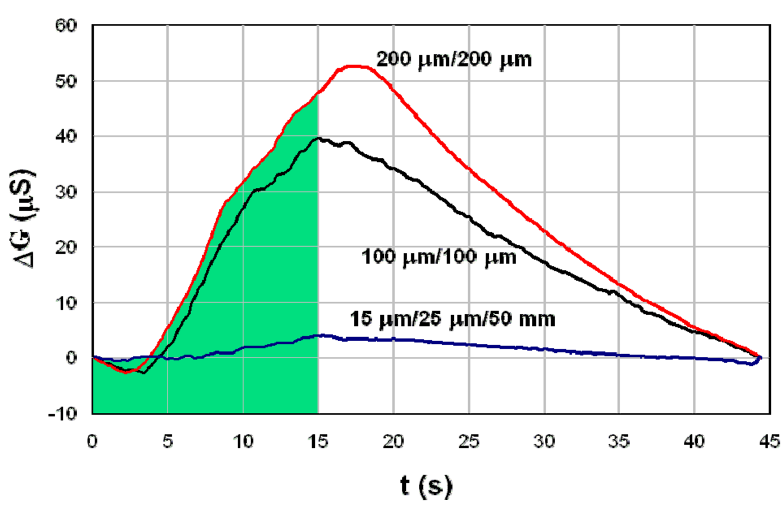


Fig. 6. Influence of size and configuration of microelectrodes on the output signals

We have also analyzed influence of sweat hydration on the output signals. For this purpose, we moistened the skin surface with NaCl solution of concentration, which was several times higher than the normal concentration of human sweat (0.3 – 0.8 %). The measurements were done using 100 μm/100 μm, 200 μm/ 200 μm and 400 μm/400 μm microelectrodes. Measured stress response was very small in case of the non-symmetric microelectrodes. Measurements showed that sweat hydration causes a noise-like influence on the signal, which gets more observable as the size of microelectrodes is smaller. Therefore, reading out the measured signal becomes more difficult (Fig. 7). Finally, one can say that sweat hydration increases the skin conductivity and brings noise to the output. Based on the experiment results, we can conclude that the skin hydration (sweat) has less influence on the output signal that might by expected, which is in correlation with the theory saying that sweat is not dominant for the current flow across the skin (Weis et. al, 1995).

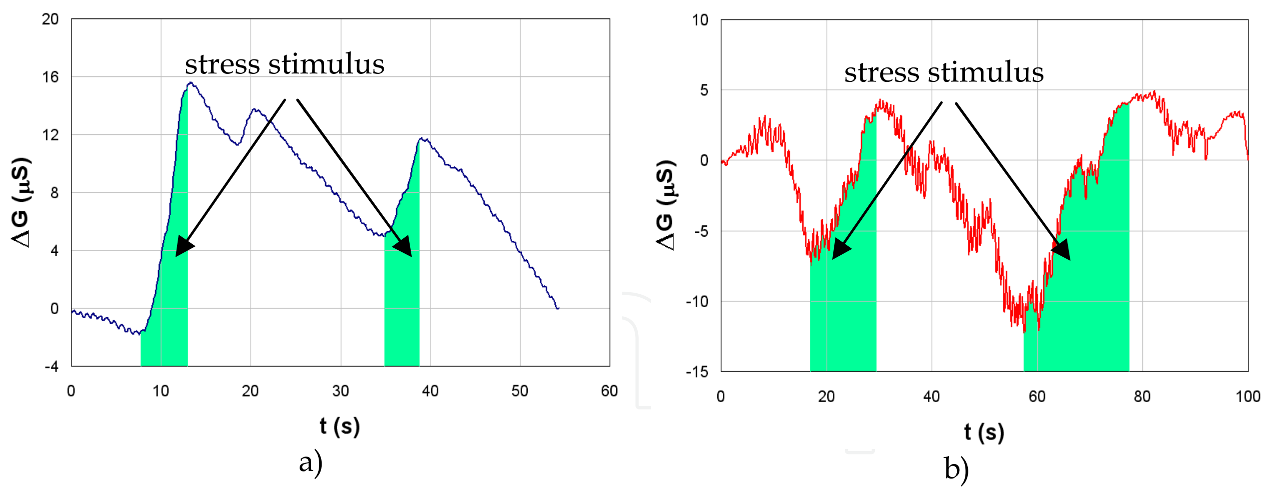


Fig. 7. Influence of sweat hydration on output signal of 200 µm/ 200 µm IDA microelectrodes: a) dry skin (G_0 = 1.22 mS), b) sweat skin (G_0 = 2.14 mS)

Finally, a comparison of our microelectrodes-based galvanic skin response (GSR) method to the commercial macroelectrode approach (Shepher, 2007), usually used in the laboratory medical or psychological experiments, was carried out. The comparison, performed using standard psychotests, shows that the responses given by both approaches were similar. However, the microelectrode signals are observed to be more stable with a shorter response time (Fig. 8).

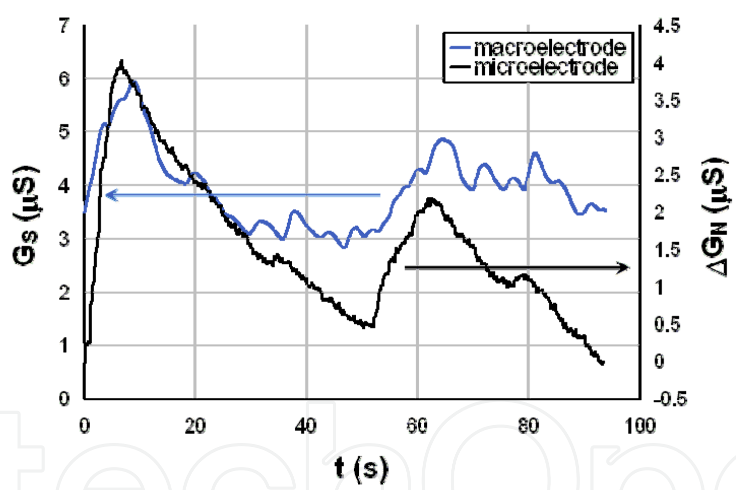


Fig. 8 Comparison of our microelectrode system and classical macroelectrode GSR method

After these experiments, the next step in our research was to develop a portable monitoring system offering sensitive and continuous measurement of PGR, with the measured data transfer to a personal computer.

3. Proposed Complex Measurement System

3.1 Measurement method

In the preliminary work, a thin film IDA microelectrodes system of different sizes and topologies has been designed, developed and realized. Then, experimental measurements of the electrodermal response (EDR) upon selected stress stimuli (invoked by different mental

tasks - standard psychotests) were performed using the developed microelectrode array. The obtained experimental results show that the optimal input signal amplitude should be selected from 1.5 V to 3V. The input signal frequency is not so critical, however, an optimal value in order of ones kHz has been proved. The most proper IDA microelectrode size is: 200μm/200μm (finger/gap ratio).
Several methods applicable to continuous measurement of the human skin impedance were analyzed first. As a result of this analysis, the auto-balancing bridge method was chosen because of few reasons (high accuracy, short time, high repeating rate of measurements, frequency and amplitude signal definition, possibility to measure both real and imaginary impedance components, controllability by a microprocessor, digital processing, etc.). Consequently, new measurement methods for both required versions of the proposed measurement equipment: a simple handy "stress- alarm" with limited features, functionality and accuracy as well as a precise laboratory-like monitoring system, have been developed. These methods are properly adjusted and modified in order to match for employing advanced available hardware accessories and peripherals. The stress-alarm method utilizes a microcontroller interface comparators for low-cost and small size. The complex laboratory measurement system is designed using precise A/D converters and with possibility to connect a lot of different integrated or external microsensors.

3.2 Developed Measurement Equipment

The complete measurement environment for continuous and non-invasive monitoring of the skin impedance has been developed. The first proposed measurement system, shown in Fig. 9, consists of the IDA microsensor described above, integrated circuit AD5933, microprocessor ADuC832, and a personal computer.

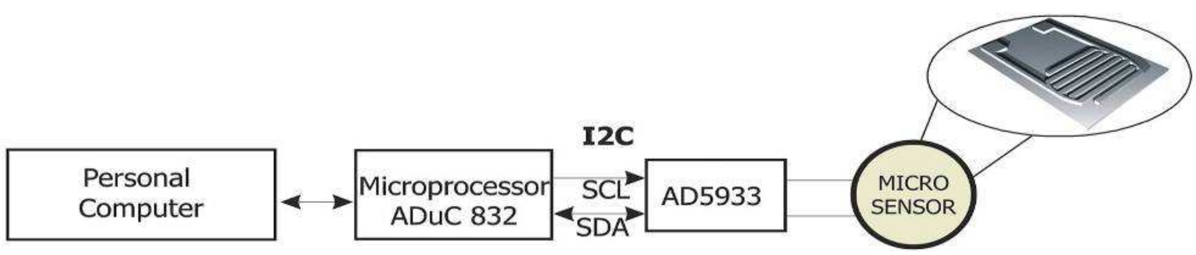
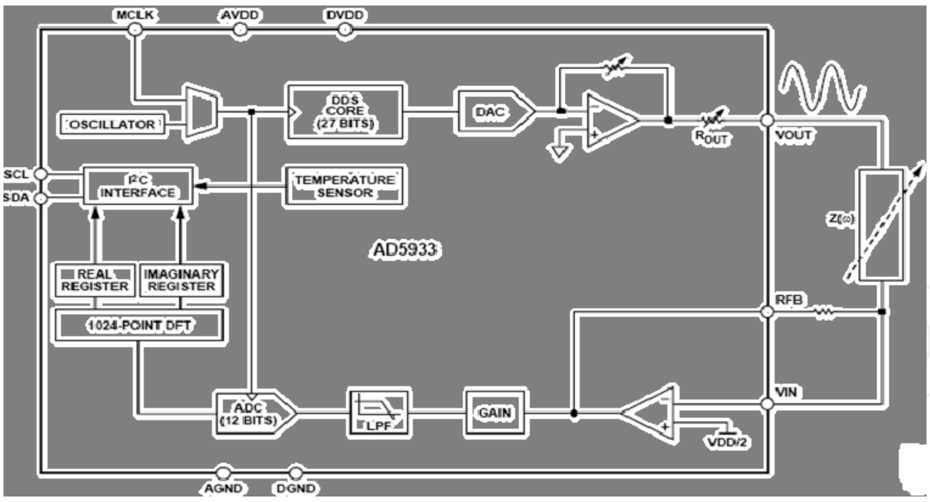


Fig. 9 Block diagram of the monitoring system using the planar IDA microelectrodes

The core of the proposed monitoring system is the integrated circuit AD5933 (Fig. 10) by Analog Devices (Analog Devices, 2007) that provides measurement of the skin impedance sensed by the developed microsensor system. The measurement process is controlled by the microprocessor ADuC832 (Analog Devices, 2007) via I2C interface. Using a serial interface RS232, the microprocessor then sends the measured data to a personal computer providing data storage. Additionally, the microcontroller also provides an initial configuration of the integrated circuit AD5933 that is needed at the measurement beginning. The configuration includes mainly setting the frequency and amplitude of the input signal used for measurement of unknown impedance. The microprocessor also controls time slots during which the measurements are performed. After the measurement in the respective time slot is done, the microprocessor reads and sends the measured data from AD5933 circuit to a PC, where data is stored and further processed.



Impedance Range	100 Ω ÷ 10 MΩ
Output Frequency Range	1k Hz ÷ 100 kHz
Internal Oscillator Frequency	16,777 MHz
Power Requirements	2,7 V ÷ 5,5 V
Total System Accuracy	0,5 %
Temperature range	-40 °C ÷ 125 °C

a) b)

Fig. 10 Integrated circuit AD5933 and its main specifications (Analog Devices, 2007)

The AD5933 circuit is composed of the following parts: an input signal generator, a 12-bit A/D converter, a DFT (Discrete Fourier Transform) circuit, a thermal sensor, and I2C interface. The generator provides a sine wave input signal of certain frequency and amplitude at the output VOUT. Unknown impedance is connected between VOUT and VIN terminals. Thus, the magnitude and phase of the current flowing through a load depend on its impedance. This current is then transformed to voltage that is converted into a digital signal by the D/A converter. Finally, the DFT circuit provides discrete Fourier transform of the converted signal. As a result, values of real and imaginary parts of a loaded admittance are measured. Photograph of the whole printed circuit board (PCB) of the monitoring system is shown in Fig. 11.

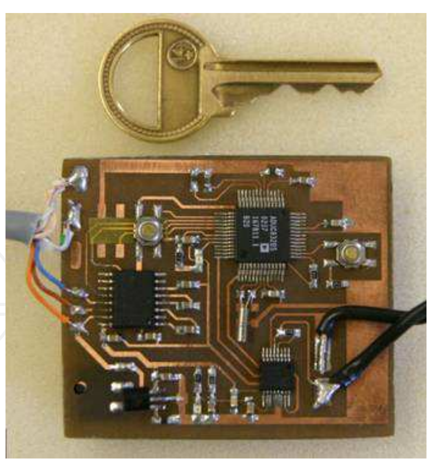


Fig. 11 PCB of the realized monitoring system.

The PCB has been realized on FR4 board by SMT technology with minimum strip width of 0.3 mm and minimum clearance width of 0.35 mm. The PCB is functionally divided into four parts: a voltage source, interface RS232, microprocessor ADuC832 and measurement module AD5933, and the power supply part providing two separate supply voltages for analog and digital parts. This is needed for the elimination of unwanted effects (e.g.

interference, noise, leakage, disturbance, etc.). Two SMD buttons ensure the software upload and microprocessor reset. The total size of the designed PCB is 50 × 60 mm.

3.3 Developed graphical user environment.

A graphical user interface (GUI), developed in C++ under Windows XP platform, provides both the necessary calibration of the measurement as well as storage, displaying, and post-processing of the measured data (real and imaginary components of the measured skin impedance). The developed software allows an easy and user-friendly control of the measurement process and data displaying and storage. From the measured data, absolute values of impedance and admittance as well as its phase are computed, and all these parameters can be displayed in several graphical and numeric modes.

3.4 Portable version of the Monitoring System

Long term biomedical monitoring plays an important role, especially, in improving diagnosis and therapeutic processes in contemporary medicine. The crucial step towards more exact and precise characterization of psychical stress influence on a monitored respondent in real life conditions is the respondent's free movement (out of a laboratory). This would enable continuous monitoring of the respondent, even during regular daily activities being carried out. From above considerations the demands on the monitoring equipment are as follows (Majer et. al, 2008):

- compact in size, low weight
- minimalization of connecting cables
- suitable sensor placement
- continual measurement ability and data storage
- low-power and battery supply

For these reasons, the developed measurement system should be small, portable and compact in size with possibilities of continuous measurement and wireless data transfer to a personal computer, eventually, storing data in memory. Moreover, this system should be able to measure other parameters (body temperature, heart pulse, etc.) for complex monitoring different psycho-physiological processes in humans. In many cases this measurement is needed to performed in plenty places of human body. The proposed complex system offers these opportunities (Fig. 12). It composed of sensing parts and the core (microcontroller with RF module, A/D, etc.) ensures measurement control, processing and transmit date. The several sensors are connected using matching circuits or wireless. One of the most important demands on the portable measurement system is the minimalization of connecting cables. Therefore, a wireless communication has been considered between the measurement equipment and the personal computer to provide necessary data transfer. The considered RF wireless communication module consists of a transmitter at the side of the measurement unit and a receiver at the PC side.

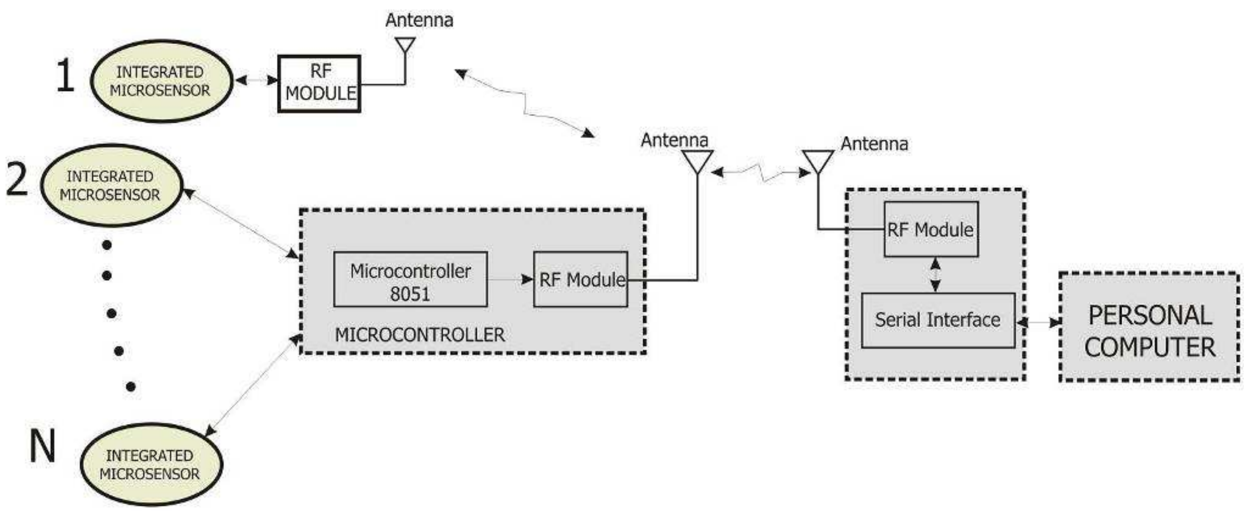


Fig. 12 A complex wireless biomedical monitoring system

Receiver module

The block diagram of the proposed receiver module, considered at the PC side, is shown in Fig. 13. There is microprocessor nRF24E1 (Nordic Semiconductors, 2008) with an integrated RF transceiver used to provide a simple, small, low-power, versatile solution at the receiver part. The RF module retrieves a signal received by the antenna and sends it to the PC via RS232 or USB interface. Microprocessor ensures control, converting, data transfer and power management.

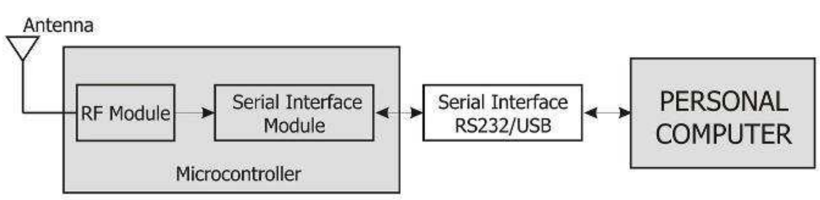


Fig. 13 Block diagram of the RF receiver communication module

Photograph of the whole printed circuit board (PCB) of the receiver module is shown in Fig. 14. The PCB has been realized on double layer FR4 board by SMT technology with minimum strip width of 0.2 mm and minimum clearance width of 0.2 mm. The total size of the developed receiver module is 11 mm × 17 mm (excluding the USB connector).

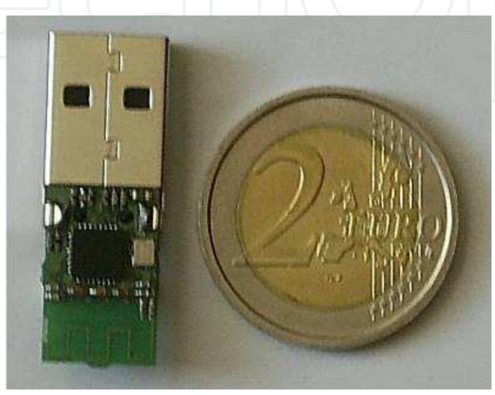


Fig. 14 PCB of a realized wireless/USB receiver module

Transmitter module

Several sensing systems with various conception and application way have been proposed. The first complete wireless measurement equipment consists of the planar microsensor, AD5933 circuit, and controlling microprocessor nRF24E1 with the RF communication module (Fig. 15). The core of the proposed portable monitoring system is again the integrated circuit AD5933 that provides measurement of the human skin impedance sensed by the developed microsensor. The measurement process is controlled by the microprocessor nRF24E1 via I2C interface. Using the RF wireless communication interface, the microprocessor then sends the measured data to the receiver part on the PC side. Consequently, the personal computer executes data storage and data post-processing. Additionally, the microcontroller also provides an initial configuration of integrated circuit AD5933 (setting the frequency and amplitude of the input signal, measurement time slots, power management, etc).

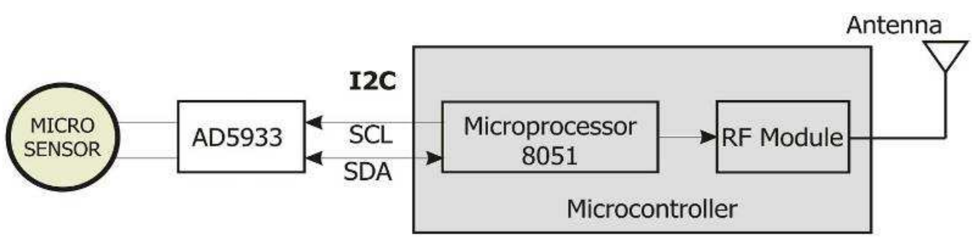


Fig. 15 Block diagram of the portable measurement system with a RF wireless transmitter communication module

The next work included mainly design and realization of the miniaturized portable version of the system with RF wireless data transfer. Thus, the integrated circuit AD5933 had to be removed, and its functions have been substituted by a microcontroller with A/D converter and RF transceiver nRF24E1 integrated within. At the same time, research and analysis of novel measurement methods, possibly applicable towards the significant miniaturization of the system and power consumption reduction (battery life time), have been performed. Two small-size monitoring equipments have been designed, applicable in both the stress alarm as well as in a complex precise laboratory measurement system.

The next step in the measurement system developing process will be led towards putting the whole system into a proper chase and making the microelectrode fixing more robust. By completing that, the laboratory or a stress alarm version of the monitoring system will be fully available.

Microcontroller nRF24E1 with RF transceiver

To reduce necessary measurement circuitry, and achieve selected features of the monitoring system (low-power, compactness, simplicity, and versatility), a modified solution of the measurement hardware has been proposed. Two microprocessors nRF24E1 (Nordic Semiconductors, 2008) with an integrated RF transceiver for the world wide 2.4 - 2.5 GHz ISM band have been employed. The RF24E1 microcontroller instruction set is compatible with the industrial standard 8051. The RF transceiver consists of a fully integrated frequency synthesizer, a power amplifier, a modulator, and two receiver units. Output power, frequency channels and other RF parameters are easily programmable by use of the RADIO register. RF current consumption is only 10.5 mA in the transmitting mode (output power -5dBm) and 18 mA in the receiving mode. The microcontroller clock is derived directly from

the crystal oscillator. The nRF24E1 allows be set into a low-power down mode under program control, and also the ADC and RF subsystems can be turned on or off under. The current consumption in this mode is typically 2 μA. The device can exit the power down mode by an external signal, by the wakeup timer if enabled or by a watchdog reset.
With respect to all the features described above, this microcontroller has been chosen for a miniaturized portable version of the measurement system with RF wireless data transfer. In this case, the whole measurement equipment consists only of several sensors with necessary electronic circuits and microcontroller with the RF transceiver module.
A well-designed PCB is necessary to achieve good RF performance. We had to keep in mind that a poor layout may lead to loss of the performance, or even functionality, if due care is not taken. A fully qualified RF-layout for the RF communication module and its surrounding components, including matching networks, has been proposed as the key issue of the proper RF design. A PCB with two layers including a ground plane is needed for the optimum performance. It is designed by surface mount technology (SMT) to achieve the best possible performance. The device sizes 0603 and 0402 have been selected. The nRF24E1 supply voltage is filtered and routed separately from the supply voltages of any digital circuitry. Long power supply lines on the PCB are avoided. All device grounds, VDD connections and VDD bypass capacitors are connected as close as possible to the nRF24E1 circuit. The PCB antenna has been chosen for better miniaturization and compactness. In this configuration, the measurement system has the access range up to several tenths of meters.

<u>Power management</u>

The power management is another crucial issue to be taken into account, since it is the best important parameter determining the operating time, and secondary, also the size of the portable equipment. There are many factors influencing on the power consumption:

1) The input signal – the measured impedance exhibits its own power consumption that depends on the input signal parameters. In Fig. 16, curves of the measured human skin impedance versus the frequency of the input signal for different electrode arrangement are shown.

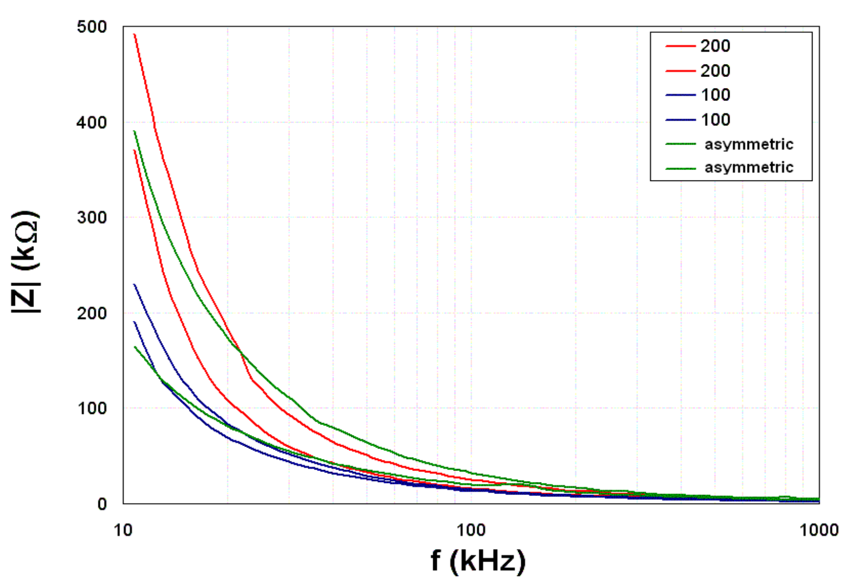


Fig. 16 Skin impedance versus the input signal frequency

The impedance value of skin is decreased as the input signal frequency arises. Other parameters influencing the power consumption are: actual value of the measured impedance, the input signal amplitude (possible range 1 - 3V), settling time after input signal is applied, sensing time of the measurement, and the frequency of sensing/sampling.

2) Power consumption of the measurement equipment – this portion is reduced by a simple design of the measurement system, and a proper power management of microcontrollers (standby mode, power down mode, etc.).

The total average power consumption is in order of ones up to tenths mW and it depends on the factors mentioned above.

3.5 Evaluation of monitoring system

Finally, experimental measurements and evaluation of the developed monitoring system have been carried out. Comparison of the results obtained by a standard HP 4284A laboratory bridge instrument measurement (G_1 - admittance) to the data measured by the developed monitoring system (G_2 and Φ - admittance amplitude and phase, respectively) is shown in Fig. 17. This measurements have been preformed in order verify realized system. The achieved results show that the developed monitoring system is sensitive to same applied stress stimuli. Moreover, the phase of the skin admittance may offer more sensitive monitoring of psycho-galvanic response than only a simple impedance amplitude sensing, since the phase reflects the admittance changes in much more significant way (high peaks in the lowest waveform).

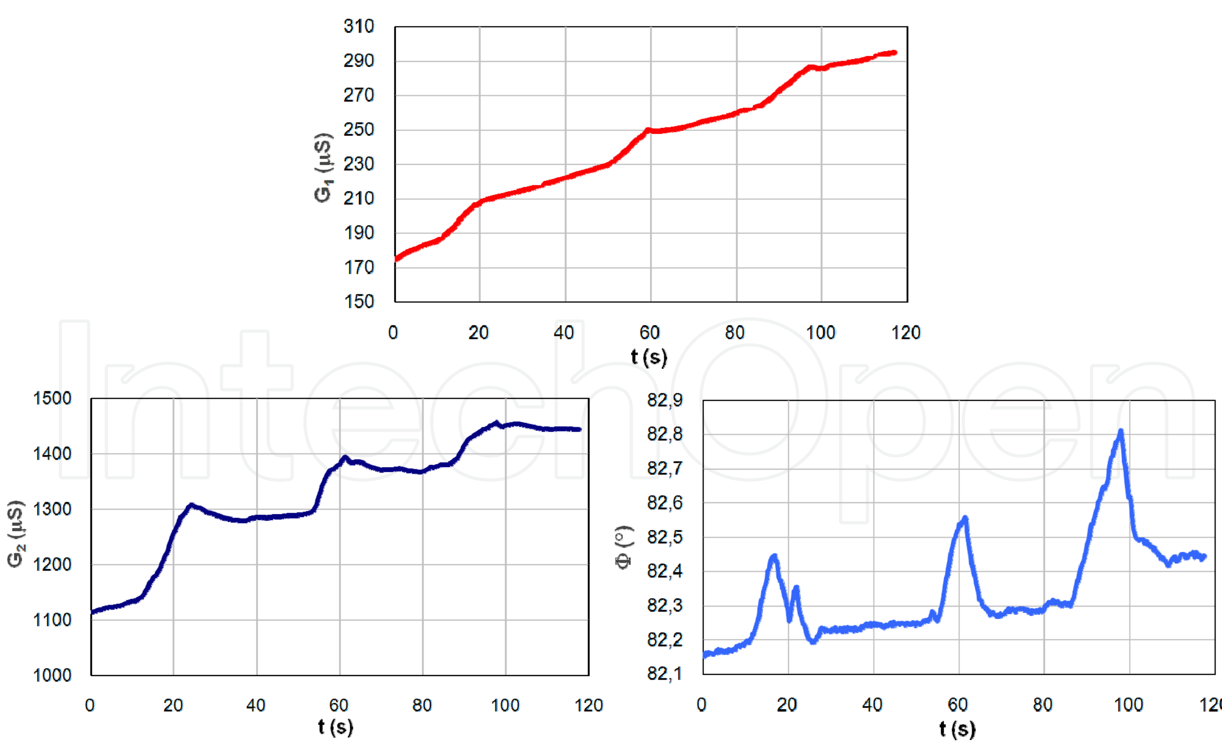


Fig. 17 Response of standard bridge instrument (G_1) versus developed system (G_2, □) in time

As the last experiment (Figure 18), our microelectrode approach was compared to a classical macroelectrodes GSR method (used on Faculty of Philosophy, Comenius University) (Shepher, 2007). The waveforms are as follows: *G3* – skin admittance obtained by macroelectrodes, *G2* and *Φ*- admittance and phase measured by the microelectrode approach. Certainly, for each method the input signal parameters were set in a proper way. In both cases, the physiological response has been evoked by the same stress stimuli. The standard psycho-tests performed have showed that the response signals obtained from both methods match and the microelectrode signals are more stable, accurate and with shorter time respond.

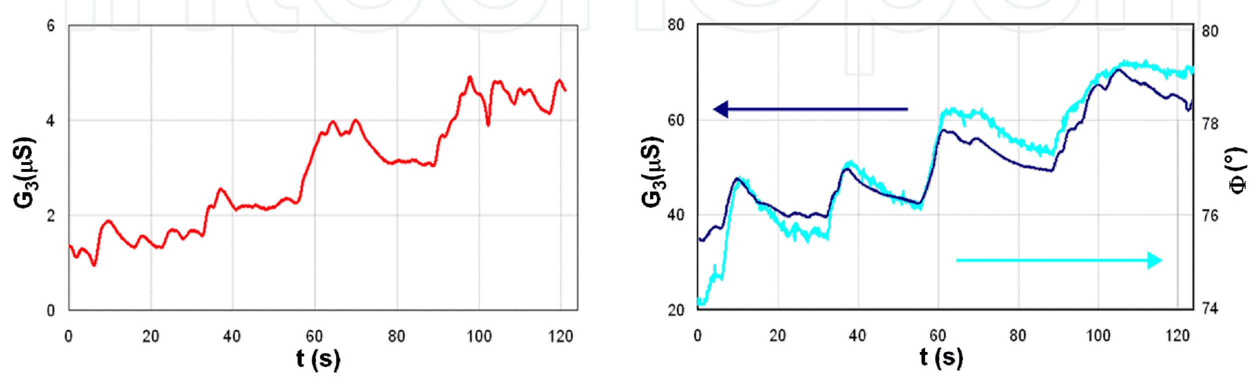
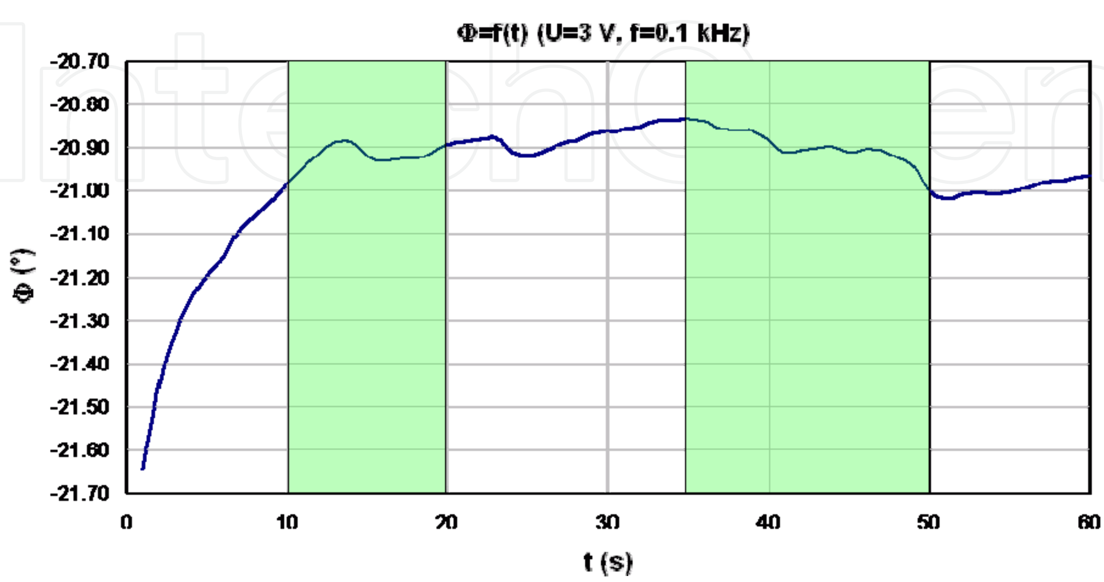


Fig. 18. Response of standard macroelectrode (G_3) versus microelectrode approach (G_2, □) in time

Fig. 19 shows curves of the measured skin admittance phase as an important factor for definition of the input signal with respect to the power consumption (stress influence periods marked by highlighted areas). It can be observed that the absolute value of the impedance decreases as the input signal frequency gets higher. Therefore, a certain trade-off between those two parameters is necessary. Moreover, the different character of the measured phase at different frequencies of the input signal has been observed, (for higher frequency – the phase decreases, for lower frequency – the phase has the increasing character).



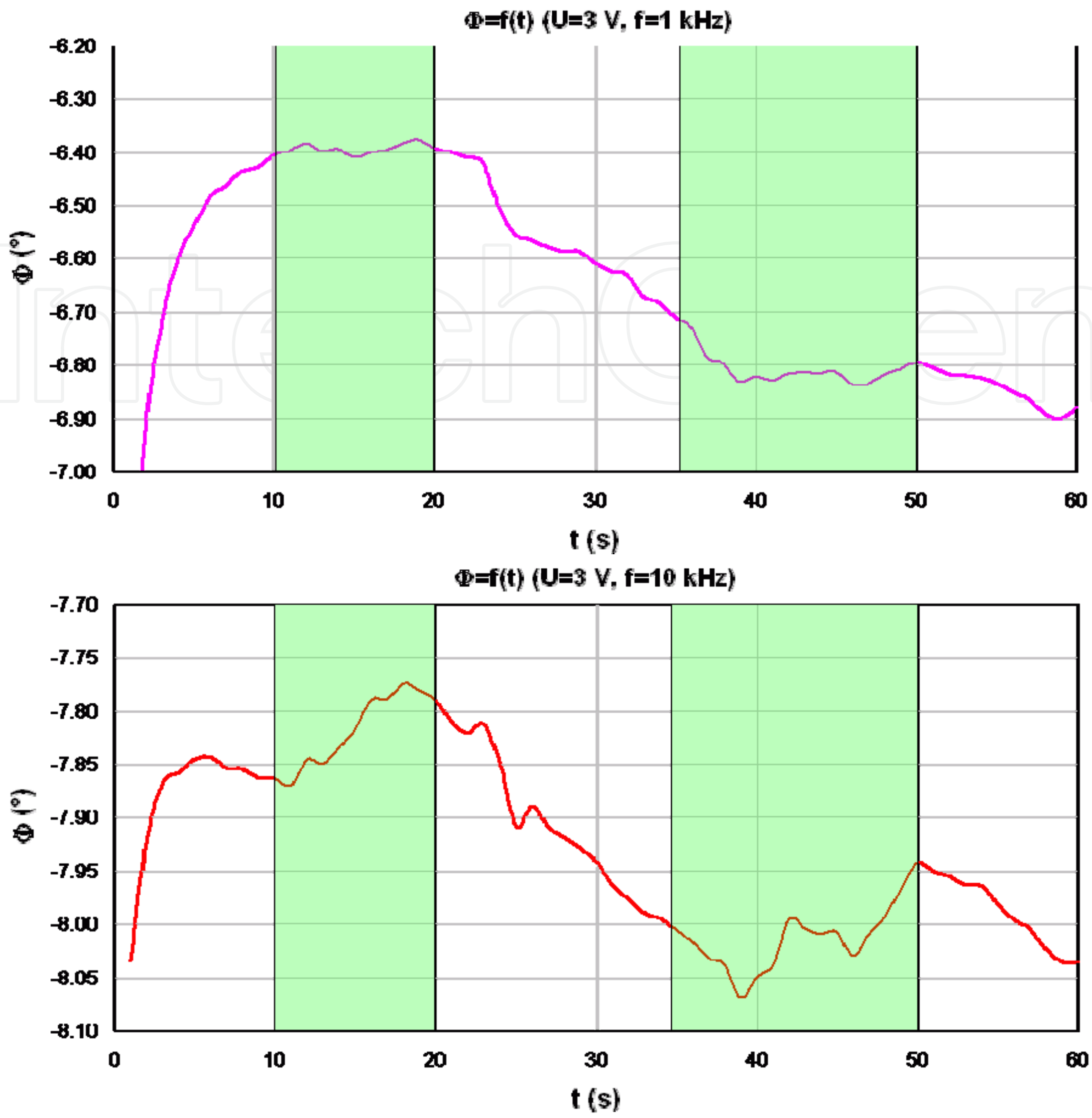


Fig. 19 Changes of admittance phase (its values and time development) at different frequencies of input signal

4. Psychological experiment

Additionally, further measurements of several selected physiological parameters, performed on a number of respondents, have been carried out as the next step of our research. The respondents were burdened by tests with various difficulties in order to invoke the proper stress conditions and achieve valuable results.

Psychological tests were focused on short-time stress situations with a large scale of tasks, which had the proband to perform simultaneously during the test. The tasks were chosen in the way to charge the same part of human brain in order to amplify the stress situation.

4.1 Psychological aspect of "Psychotest"

The presented experiment "Psychotest" was done in the laboratory of cognitive processes at Department of Psychology, Comenius University in cooperation with Department of

Microelectronics, Slovak University of Technology. This experiment has been done on a group of 35 probands in age between 19 and 30 years. The “Psychotest” was oriented to evoke short-timed stress stimulus with wide range of tasks, which a proband had to solve simultaneously. The software program Neorop II., test of distraction stress “SPEED” was used. All tasks are chosen in the way to charge the same part of human brain – in order to increase the overall stress activation (Brezina, 2007). Two types of “Psychotest” were used: “Test A” with lower stress activation and “Test B” with higher stress activation. Both tests are separated in two parts with different periodicities of tasks.

4.2 Measurement set-up

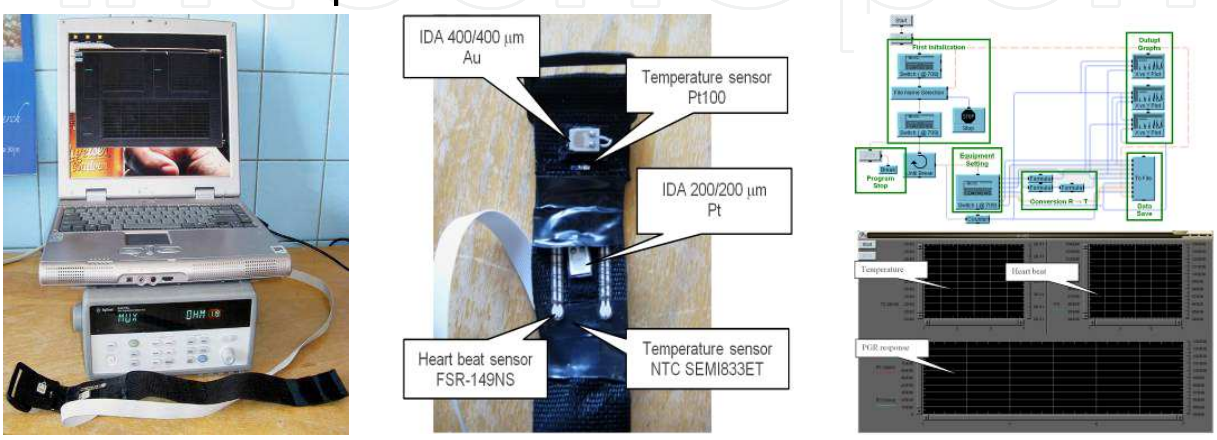


Fig. 20 Microelectrodes configuration detail

During the “Psychotest”, psychogalvanic response (skin conductivity) and body temperature, sensed on the top as well as the bottom part of left hand wrist, have been monitored and analyzed. Pressure sensors FSR-149NS for heart pulse monitoring have been used too, but with unsatisfactory results. The measurements were done using 20-channel Switch Agilent 34970A millimetre controlled over data bus line GPIB by programmed software "PGR - switch" in the measurement environment Agilent VEE (Fig. 20).

Two different IDA microelectrodes and two types of the temperature microsensors have been employed. The skin conductance was monitored using a 400 / 400 μm (finger / gap dimensions) gold IDA microelectrodes (placed on the top part of wrist) and a 200 / 200 μm platinum IDA (bottom part of wrist). The body temperature was measured using NTC SEMI833ET (top) and Pt100 (bottom) microsensor.

We have minimized the drift effect like in chap. 2.4 and the final normalized signal ΔG corresponds to the difference between actual skin conductance and the approximation function: $\Delta G(\mathrm{t})=G(\mathrm{t})-G_{\mathrm{N}}(\mathrm{t})$. The goal is to make psychogalvanic response easy readable. As the initial step of the experiment, we have tested a reference proband out of psychological test. In the first third of the experiment, the proband was without any activation. In the second third, he was very slightly activated through talking, and finally, in the last third, he was activated only via continual computer test of distraction stress “SPEED”. As shown in Fig. 21, a consistent decreasing slope of temperature response and the respective consistent steep increase of the skin conductance response can be observed. Fig. 21 shows total variations of the measured parameters in different activations levels of the experiment.

An example of typical "Psychotest" responses is shown in Fig. 22, where one can observe variations in activation (stress stimulus) and relaxation (adaptation) phases.

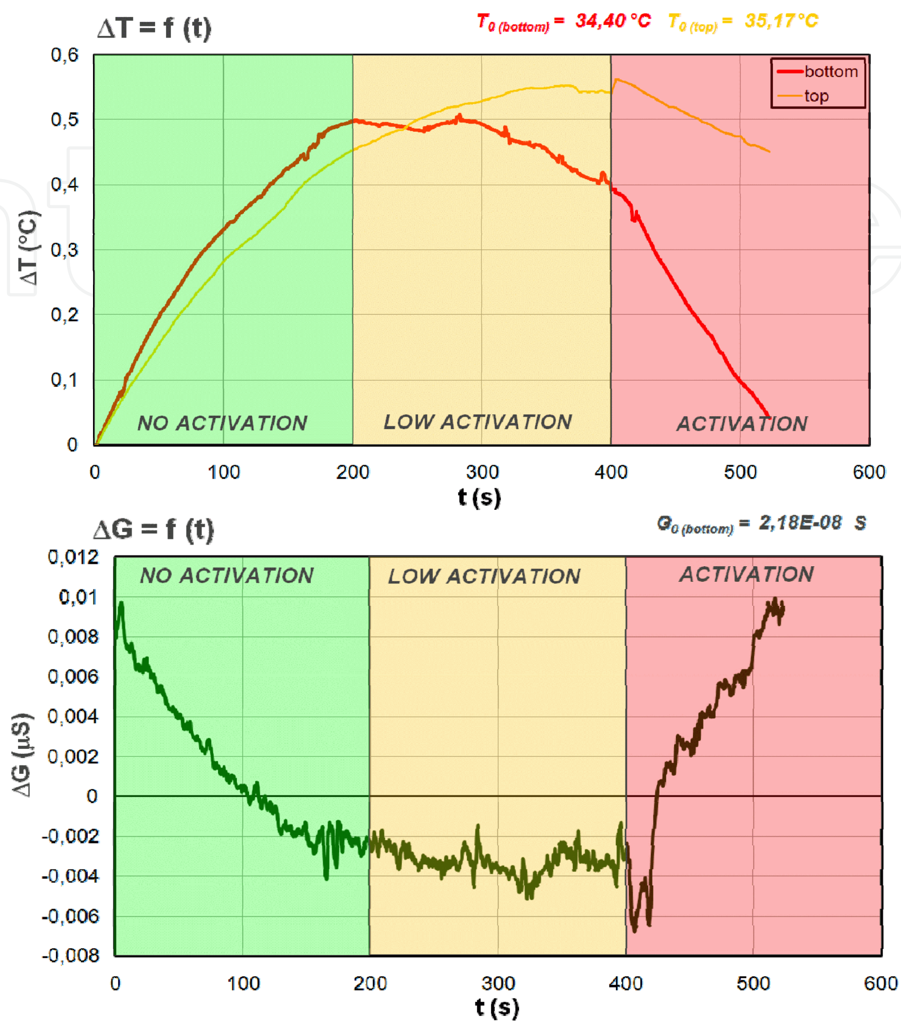
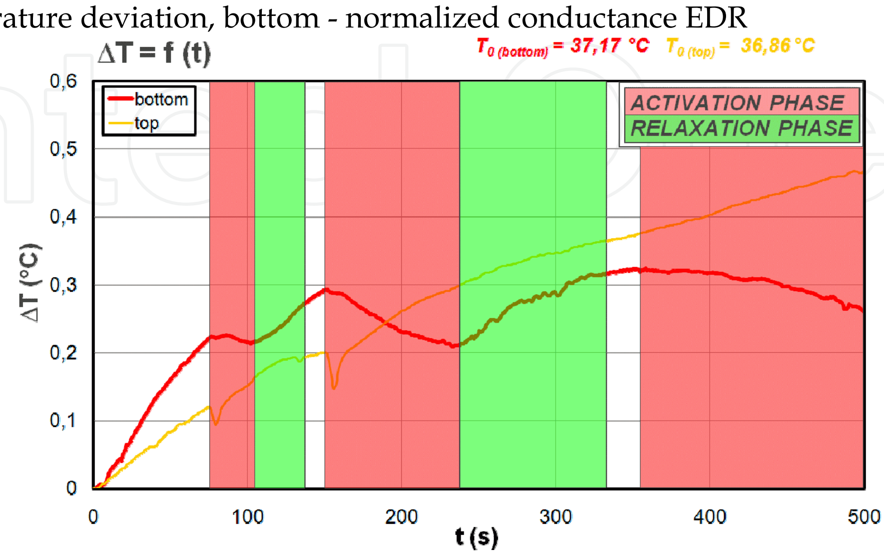


Fig. 21 Physiological responses on different activation levels:
Top - temperature deviation, bottom - normalized conductance EDR



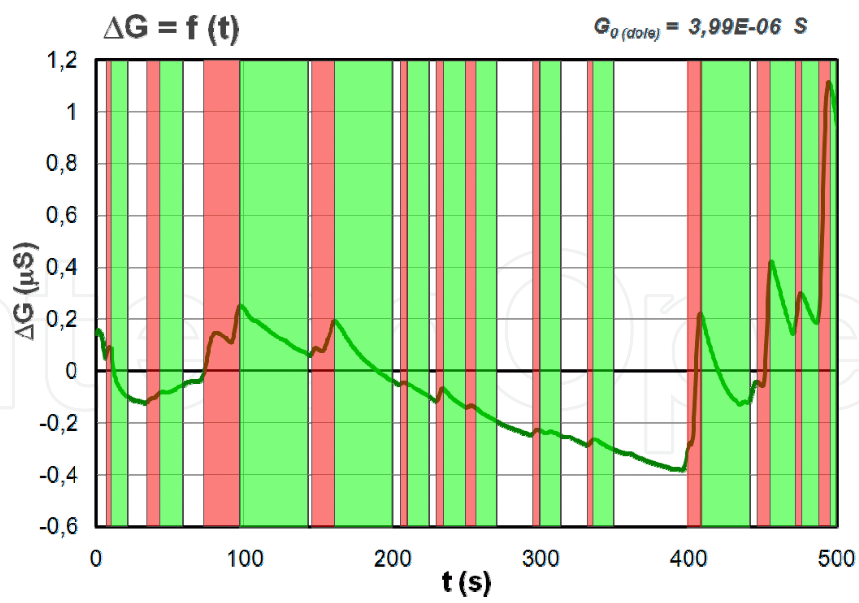
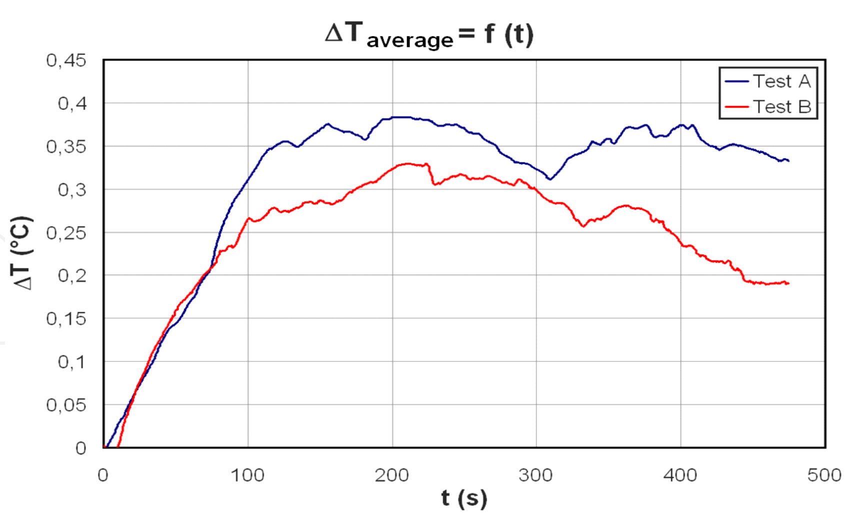


Fig. 22 Activation and relaxation phases:
Top - temperature deviation, bottom - normalized conductance EDR

Finally, the measured parameters of both tests: Test A - „lower stress activation“ versus Test B - „higher stress activation“ have been compared (Fig. 23). The presented results represent the average from all the entered probands' responses. One can observe a difference between the measured physiological variables depending on the stress activation level. Decreasing average temperature obtained within "Test B" probands is more significant as that of "Test A". Average increasing conductance of "Test B" probands is substantial as that of "Test A". This is due to the fact that "Test B" probands were psychologically more activated, and the resulting stress is rather substantial.



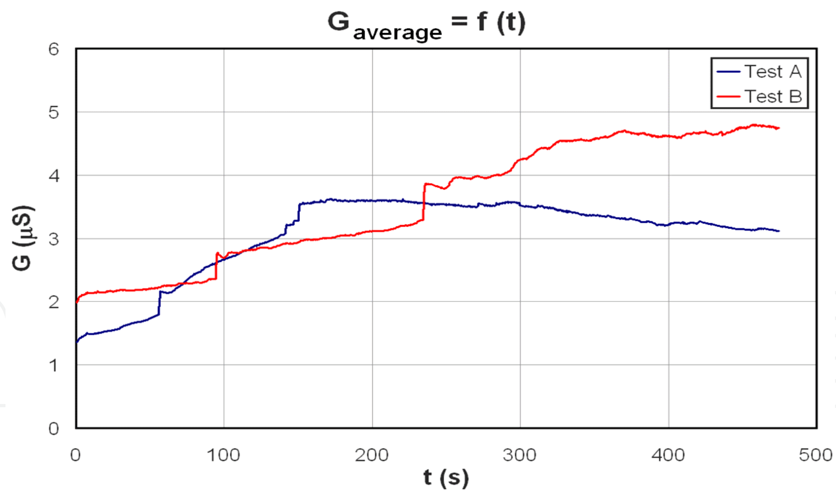


Fig. 23 Comparison of average EDR (Test A versus Test B)
Top - temperature deviation, bottom - conductance EDR (wrist bottom)

4.3 Achieved results

Psychological test experiments have been performed on a group of probands and two basic human skin parameters were sensed and analyzed. Psychological activation results to the temperature decreasing of human body and at the same time to the increasing of skin conductivity (Fig. 21). The particular parameter amplitude depends on the stress activation level and it is individual for each proband (Fig 22). Conductance EDR to the stress activation (stimulus) is shorter (2 - 5 seconds) in comparison to temperature response (20 – 120 seconds). The temperature response shows more integral character (Fig. 22). Activation phase of the parameter responses is followed by relaxation phase. The amplitudes values are getting back to their initial values. Skin conductance relaxation phases are 2 – 4 times longer than activation phases. Temperature relaxation phases are equal to the activation phases (Fig 22). During the performed "Psychotest", majority of probands exhibit increase of immunity after repeatable stress activations, and therefore, corresponding EDR amplitudes were decreasing. Both temperature and conductance sensors were more sensitive to psycho-physiological activations if placed on the bottom part of wrist compared to top part.

5. Optical reflectance of human skin

The dermis (deep inner layer of the skin) is heavily permeated with blood vessels containing haemoglobin. Haemoglobin is a protein contained in the red blood cells (95% of the dry mass of red cells). Haemoglobin binds very easily to oxygen, making it the ideal "vehicle" for the transportation of oxygen from the lungs to the tissue. Haemoglobin has an unique light absorption spectrum with characteristic absorption bands at 420 nm, and in the 545 - 575 nm wave length range, where the "W" pattern can be observed (Fig. 24b) (Angelopoulou, 1999), (Gerasimov, 2003).
These absorption bands occur only when haemoglobin is bound to oxygen. De-oxygenated haemoglobin exhibits rather shifted absorption bands, and the distinct pattern is no longer present (Angelopoulou, 1999).
Therefore, using optical (light reflectance) measurements, the quantity and oxygenation of haemoglobin in top layers of the human skin can be easily monitored that might offer

another very important input factor in monitoring some psychosomatic processes. The advantage of the optical method is also in the contactless manner of monitoring, which is independent on the contact quality variations due to the possible physical activity of the respondent during testing.

5.1 Optical measurements

In the experiment for the optical method evaluation (Fig. 24, Fig. 25), the optical spectrum (reflectance), measured under certain stress stimulus in time domain again on the wrist bottom, has been analyzed. Experimental measurements were done using optical spectrometer AvaSpec 2048 and relevant optical sources (halogen + deuton light) (Fig 24a). The light was emitted and measured using an optical fibre at distance about 10 mm over the skin surface. Maximum changes of the skin reflectance occur in around 540 nm and 576 nm wavelengths, which perfect corresponds to the haemoglobin absorbance ranges (Angelopoulou, 1999), (Gerasimov, 2003) (Fig. 24b).

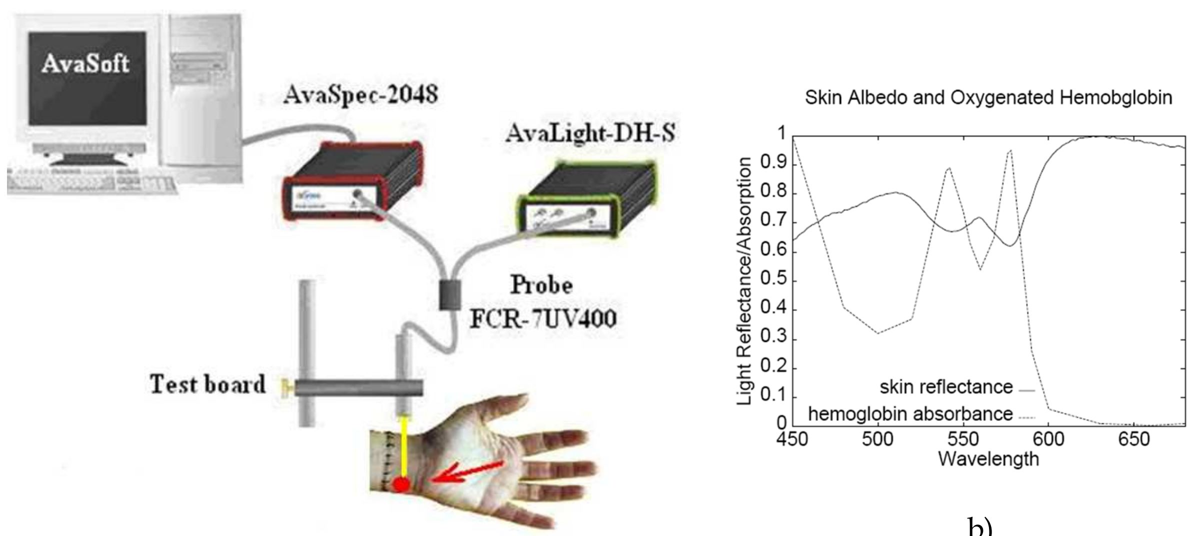


Fig. 24 Optical setup: a) equipment, b) haemoglobin absorbance and skin reflectance

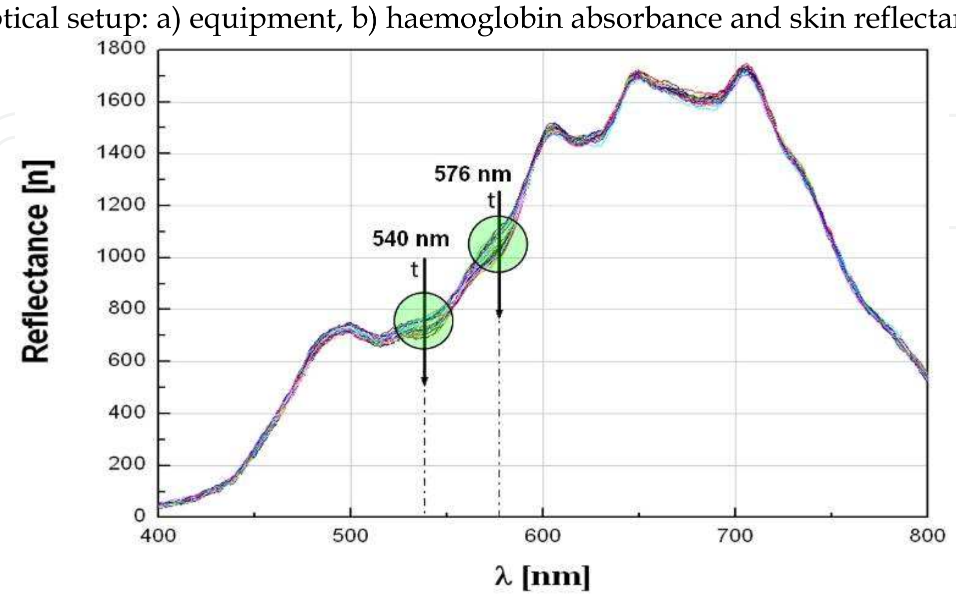
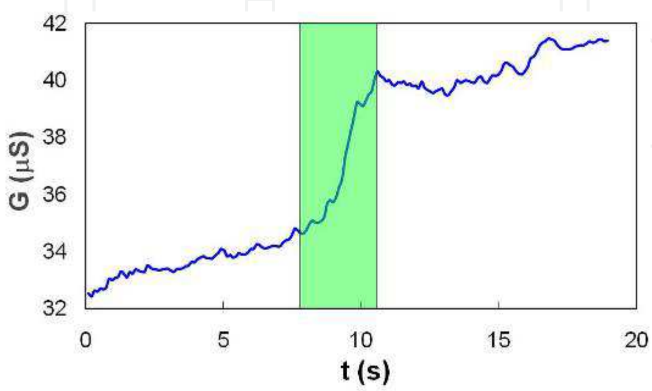


Fig. 25 Time response of skin optical spectrum to the stress stimulus

5.2 Electrical versus optical measurement

In the last experiment, both the electrical and optical measurements of psychosomatic processes are compared (Fig. 26). All the measurement were performed simultaneously, and evoked by the same stress stimuli (highlighted areas). The comparison shows rather perfect correlation of both methods in terms of the stress response monitoring ability and sensitivity.



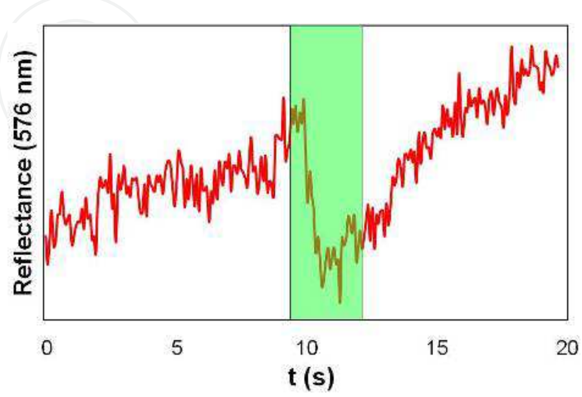


Fig. 26 Electro-optical comparison: a) electrical, b) optical

6. Conclusions

The achieved results have proven that the dedicated developed microelectrodes are able to sense the EDR in a very sensitive and fast way. Moreover, experiments, prior to the work presented here, prove that the achieved accuracy, input signal voltage and frequency ranges are suitable not only from biomedical monitoring point of view but also from the measurement system integration and miniaturization requirements.

A non-invasive portable measurement system for the reliable and precise stress detection, based on the psycho-galvanic reflex monitoring, has been designed and developed. The system is based on an auto-balancing bridge measurement method offering digital processing and displaying of the measured data in the developed GUI operating under Windows XP platform. The measurement equipment uses microelectrodes developed for this purpose, and utilizes microcontrollers containing RF wireless communication modules to transfer data between the measurement unit and a personal computer. The complete measurement system has been designed with respect to the system accuracy, sensitivity and power management. Interesting outcome has been observed – the psychogalvanic reflex might be much more accurately sensed by the skin admittance phase, since this parameter reflects the human skin conductivity changes more significantly. The achieved measurement accuracy, input signal voltage and frequency ranges are suitable not only from human biomedical monitoring point of view but also from the measurement system integration and miniaturization requirements.

Additionally, the proposed system is versatile and flexible, easy to be extended by other sensors' types (integrated or external with wireless communication feature) or different measurement approaches, which enables measurement of other physiologic parameters (e.g. body temperature, blood pressure, heart beat, etc.) There is no doubt that the developed measurement equipment offers new opportunities towards non-invasive wireless system for continuous biomedical monitoring, applicable in diverse domains, such as clinic psychology, medicine or other everyday life areas. All these features and properties make the developed biomedical monitoring system very helpful also from the life quality enhancement point of view.

We have also analyzed a new approach to measurement of the psycho-galvanic reflex by skin optical reflectance method. This method seems to be very suitable for monitoring of the quantity and oxygenation of haemoglobin in top layers of the human skin, and implementation to the final integrated setup.

Motivated by the promising results achieved so far, the research will go on by the next step that is integration of the whole monitoring system into a single chip, working under low-voltage and low-power conditions that would meet basic requirements for modern portable wireless biomonitoring equipment.

7. Acknowledgement

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